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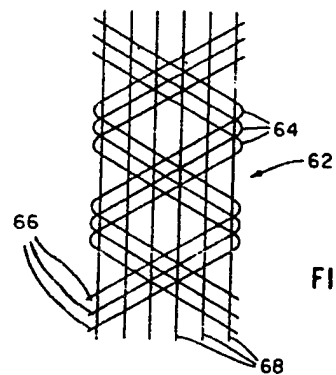
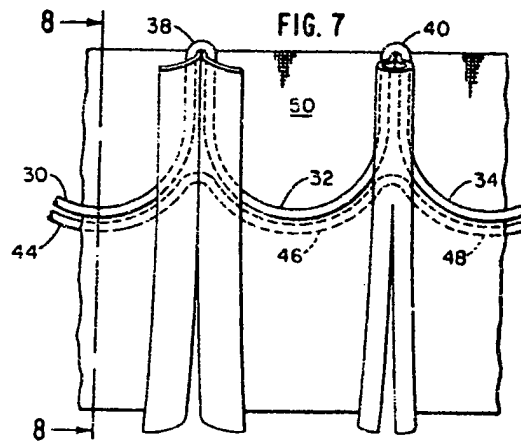
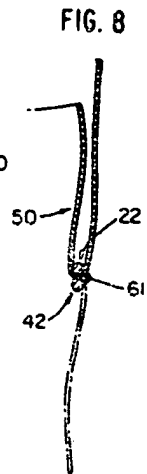
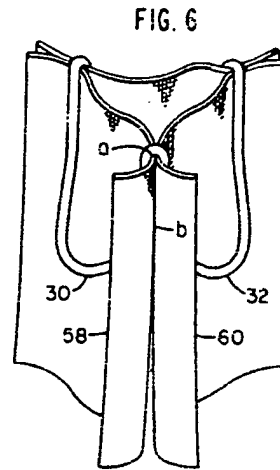
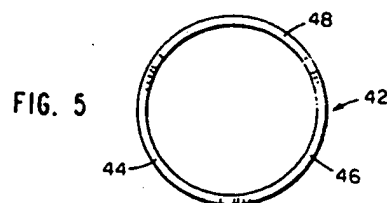
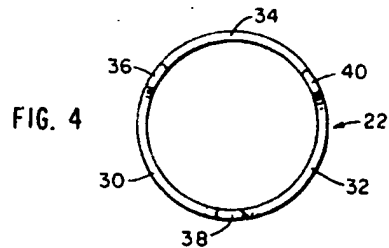
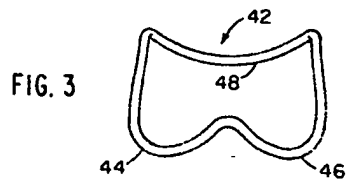
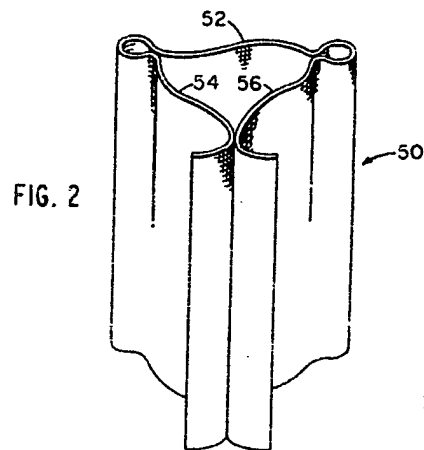
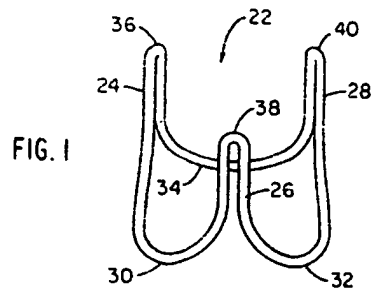
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(54) **Elastomeric sheet materials for heart valve and other prosthetic implants**

(57) Flexible and extensible sheet materials for cardiovascular and other prosthetic implants are fabricated of synthetic elastomers and have at least one fibrous reticular face. These materials are characterised by molecular configurations that produce a nonlinear tensile modulus characteristic similar in shape to that of natural tissue. Selected elastomers, when formed as sheet material of appropriate geometrical configuration, promote the formation of natural tissue thereon which is appropriate for the mechanical characteristics desired.

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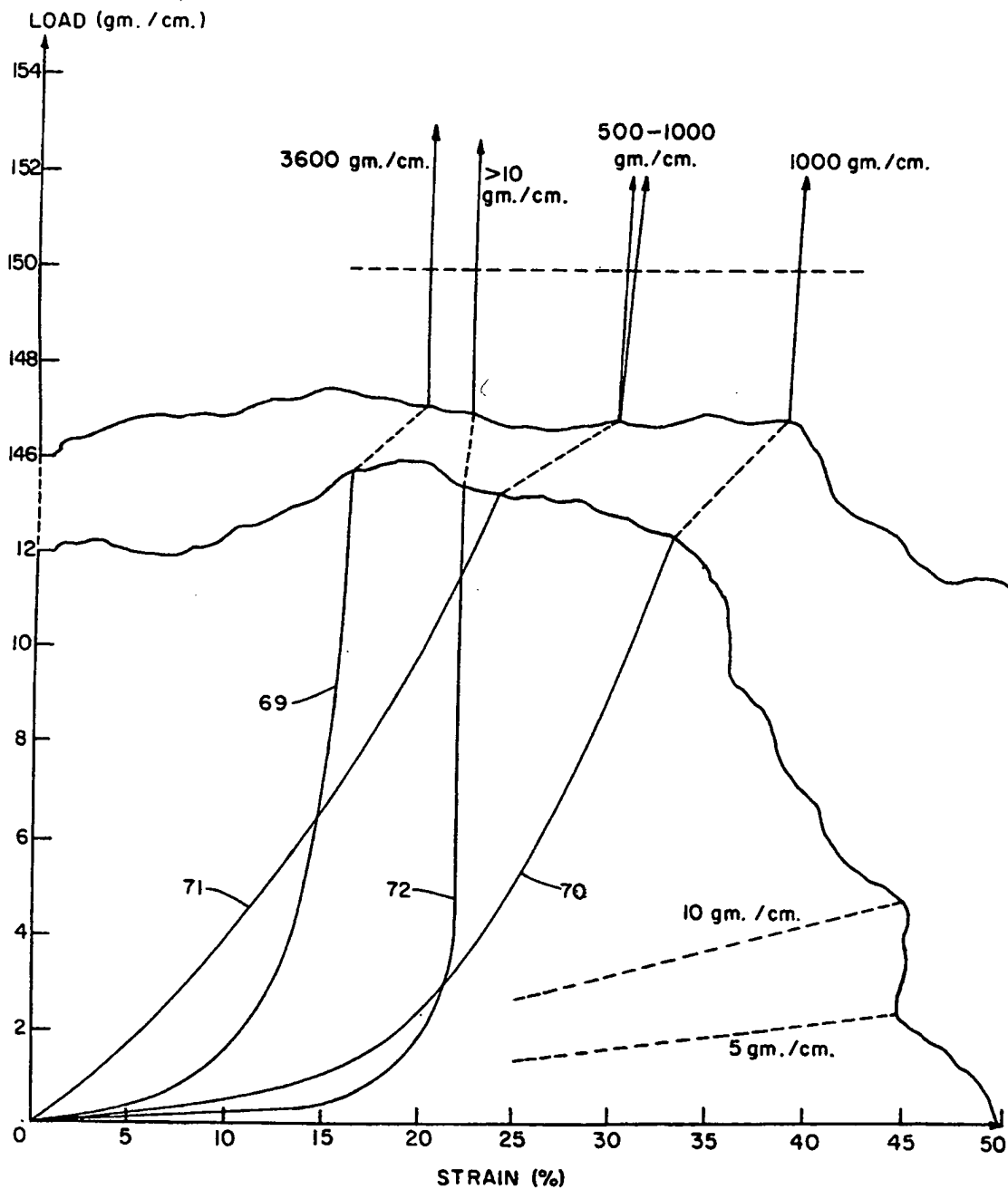


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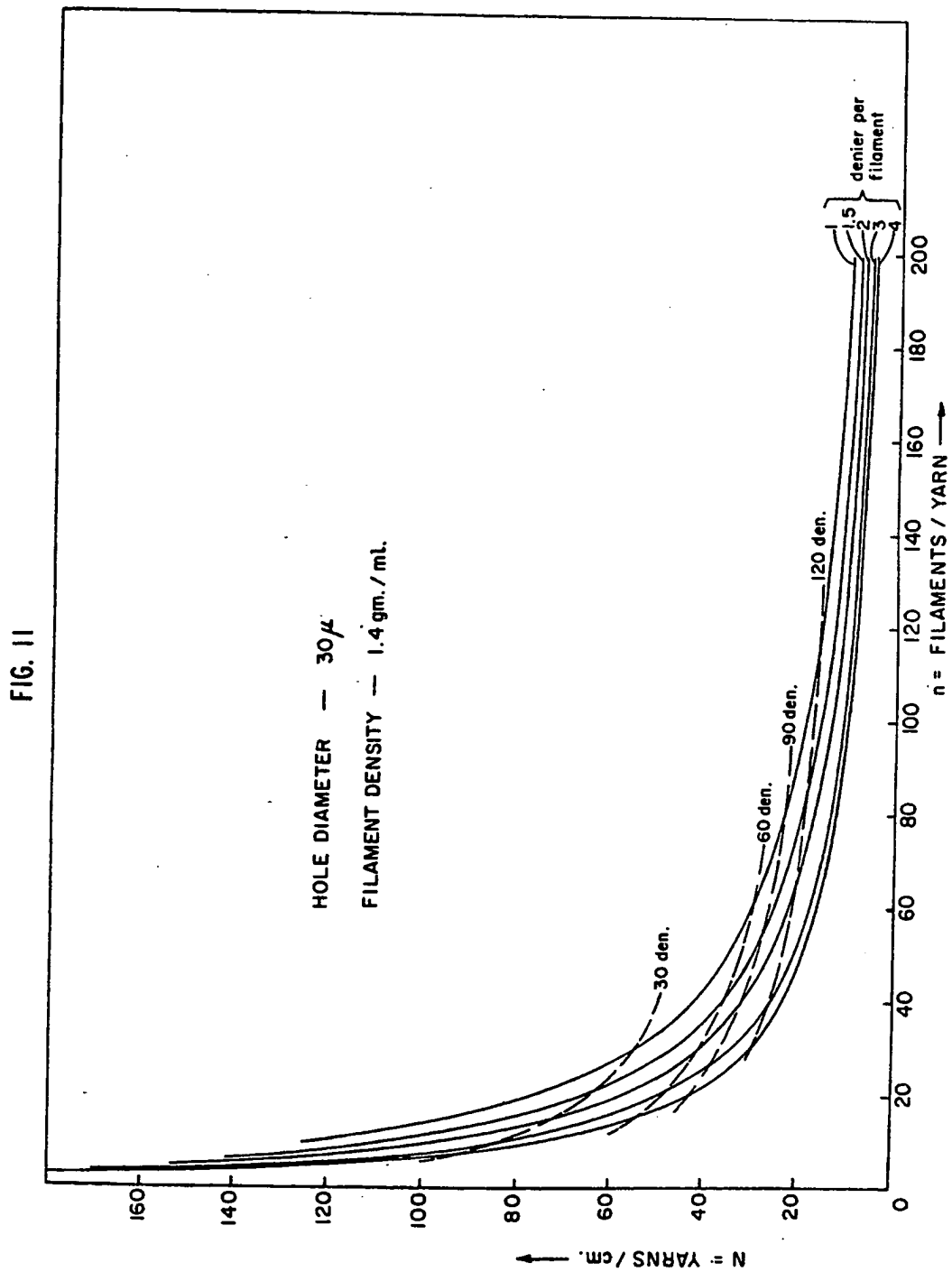
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FIG. 10



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## SPECIFICATION

## Elastomeric sheet materials for heart valve and other prosthetic implants

5 This invention relates to sheet materials for vascular, heart valve and other prosthetic implants. More particularly, it relates to such sheets fabricated of synthetic elastomeric materials.

10 This invention is directed to providing synthetic sheet materials having properties in use that closely approximate those of the natural tissues that they replace. For an understanding of these properties the human aortic valve may be taken as an example,

15 in that it typifies properties that are required for other types of implants as well, such as vascular implants. This valve is of the leaflet type having thin flexible membranes with a face to face thickness of about 0.06 cm that open 70 to 90 degrees from horizontal into the surrounding blood vessel (ascending thoracic aorta), and form three contiguous pouches held in close and leak-proof mutual contact by the pressure of the blood when in the closed configuration. Thus the membranes cause a minimum of disturbance to the flowing blood when in the open configuration, but move quickly when the blood pressure reverses (changes sign) to prevent regurgitation. A number of properties of the human valve may be identified, and these comprise the more specific objects of the present invention.

30 A first characteristic of the human aortic valve is that its response time is minimal. Thus the present invention provides a synthetic sheet material that has a low resistance to motion of the leaflets in terms of both the inertial and elastic components of such resistance. In general terms, this is achieved by making the leaflets as light in weight and as flexible as possible consistent with the other mechanical requirements of the valve discussed below. This will enable the valve to pass quickly from the fully open state to the fully closed state when the pressure differential changes sign, leading to reduced energy losses in the flowing blood, and a minimum of undesirable regurgitation.

40 A second property of the human aortic valve is the effectiveness of the seal. Observations of this function reveal that the sealing of the valve is accomplished by the intimate conjugation of regions (referred to as the coaptation zone) close to the free edges of the valve leaflets. The effectiveness of the seal depends upon the degree of compliance of the leaflet in directions both parallel to its plane and transverse to that plane. The high transverse compliance allows the two contacting surfaces to form a more intimate conformal fitting at the coaptation zone, and the high in-plane compliance insures that the coaptation zones is sufficiently large to make an effective seal. Experience with synthetic valve leaflet implants has shown that these two forms of compliance are influenced by different factors. When such implants are employed, natural tissue is deposited or formed on the leaflet, and the nature of this natural tissue depends upon the nature and geometry of the synthetic material in use. The transverse compliance is controlled to a large extent by

the mechanical properties of this tissue. In contrast, the in-plane compliance is controlled directly by the mechanical properties of the substrate synthetic material. Therefore, for suitable in-plane compliance

70 this invention provides synthetic materials having longitudinal (i.e. tensile) compliance similar to that of the human leaflet. In addition, for adequate transverse compliance, additional criteria must be met with respect to the promotion of suitable natural tissue on the surface of the implant fabric.

75 A third property of the human aortic valve is the provision of adequate load bearing capacity. For a more detailed understanding of this property, a distinction may be made between the tensile compliance in the circumferential direction of the cusp, that is, the in-plane direction parallel to the free edge of the valve leaflet, and the tensile compliance in the radial direction of the cusp, that is, the in-plane direction perpendicular to the free edge. In either direction, the maximum working load level may be taken for practical purposes as approximately 150 gm/cm of leaflet width. This load exists in the closed configuration of the valve during the peak operating pressure in the arterial system. These requirements for adequate load bearing capacity would appear to impose properties different from those associated with the high compliance requirements previously described. However, the paradox is resolved in nature by the marked nonlinearity of the stress-strain characteristic of the natural leaflet material. At low loads the material has an extremely low modulus, thus ensuring both rapid response and good conformal fitting, but at a particular value of elongation (typically in the 10% to 20% region) the modulus undergoes a marked increase, with the result that the natural tissue can sustain high levels of loading without excessive subsequent elongation, gross geometric distortion or rupture.

100 Quantitatively, the foregoing features may be described as follows. In the circumferential direction the natural leaflet extends very readily with increased load until an elongation of 10% to 12% is reached at a load of 1 to 2 gm/cm of leaflet width. Upon further increase in the load, the resistance to further elongation increases greatly, and at the maximum working load level of 150 gm/cm width the modulus is approximately 3600 gm/cm width, which corresponds to a modulus expressed in classical engineering units of 850 pounds per square inch (p.s.i.). In the radial direction of the cusp, the region of easy extension with increased load, continues up to approximately 20% elongation, at which the load is about two gm/cm width. Upon further increases in load the resistance to further elongation, though greater than in the initial region, is not as high as it is in the circumferential direction. The working load in this direction is not as firmly established as in the circumferential direction, but at a load of 150 gm/cm width the modulus is approximately 1000 gm/cm width (250 p.s.i.). Thus in each direction there is a transition between an initial region in which the modulus is of the order of 10 gm/cm width and the high load region in which the modulus is between 1000 and 3600 gm/cm width.

130 In view of the variations that occur in the tensile

moduli of natural heart valve and vascular tissues, it is difficult to ascribe exact significance to the absolute values of the moduli described above. However it appears that satisfactory performance of a synthetic prosthesis can be obtained if a modulus can be achieved which is within a factor of two of the natural leaflet values given above. Accordingly, it is preferred that the fabrics of this invention have a tensile modulus that is no greater than about 7200 gm/cm of leaflet width nor less than about 500 gm/cm of leaflet width up to a load of about 150 gm/cm of leaflet width.

In U.S. patent No. 4,191,218 the property of non-linearity in the natural valve leaflet material is generally described, and a synthetic leaflet material composed of multifilament polyester yarns is disclosed. To obtain the desired nonlinearity in the stress-strain characteristic, these yarns are subjected to compaction and shrinking to produce crimps in the yarns. The easy compliance of the crimped yarns at low stress levels results from the fact that the crimps are being straightened out; and the substantially higher tensile modulus exhibited at higher stresses results from the fact that the straightened yarns are being stretched.

Another property of the natural valve leaflet is its ability to maintain its original geometry and structural integrity through a large number of cycles of stress. For these purposes four times  $10^9$  cycles may be taken as the pertinent criterion. Accordingly, this invention provides a sheet material that has a rapid and near-complete recovery from applied stress. The fatigue lifetime of a material that is subjected to cyclic stress and strain is influenced by the amount of non-recovered work (hysteresis loss) that is associated with the stress cycle. If there is rapid and substantially complete elastic recovery of the material from a cyclically applied stress, a longer fatigue lifetime will generally result. Further, it is desirable to provide a sheet material in which the applied stresses do not undergo a change of direction, as such a change tends to reduce the fatigue lifetime of most materials.

In addition to providing properties similar to those of natural heart valve leaflets, synthetic leaflets, vascular implants and the like must be so structured as to promote desirable tissue overlay when implanted, and the materials used must have adequate resistance to chemical change in the implanted environment.

According to this invention, sheet materials for cardiovascular and other prosthetic implants are fabricated of synthetic elastomers. These elastomers, when formed with appropriate geometrical configuration, exhibit all of the properties of the natural heart valve tissue previously described to a higher degree than those materials hitherto used. As used herein, elastomers are defined as polymeric materials that exhibit rubber-like elasticity characterised by low modulus (high compliance) and hence considerable elongation under low loads, with ultimate elongation reaching up to 1000% in some cases, with rapid recovery toward the original, unstretched dimension on removal of the stress, and with little or no permanent deformation as a result of the imposi-

tion of the stress. In polymeric materials of high molecular weight, this type of behaviour is associated with a relatively small number of permanent crosslinks, and with rubber/glass transition temperatures considerably below the operating temperatures. The elastic extension in these materials is associated with the change in the configuration of the long polymer molecules from a random coil in the unstretched state to the extended condition. The resistance to such a change increases as the molecules become increasingly aligned with the direction of stress, and the stress-strain curve usually shows an increase in slope as the elongation is increased. This is the type of elastic response that is needed to correspond with the response of the natural valve.

There are certain differences between the elastomers of this invention and the crimped fibres described in the above mentioned copending application. In the elastomers of this invention the non-linearity in the stress-strain characteristic is produced at the molecular level; whereas, in the crimped fabrics described in said application the nonlinearity is imposed at the supramolecular, that is the fibre level, being associated with the bending deformation of the crimped fibre. The process of unbending and rebending of the crimped filaments in the yarns is associated with cyclical stress reversals. These stress reversals due to this bending do not occur in the elastomers of this invention; therefore, an improvement in mechanical fatigue characteristics and a longer useful lifetime are achieved. Moreover, the elastomers of this invention differ from the polyester of said application in the higher modulus portion of the stress-strain curve. The differences here are exhibited by the relatively more rapid and more complete recovery of the elastomers when the stress is removed.

To achieve the desired minimal inertial and elastic resistance of the synthetic sheet material to the motions introduced by the hydrodynamic flow conditions, this invention involves the use of materials having a relatively low mass per unit area of the leaflet materials. Also, the second moment of area of the cross-section and the bending modulus of the material are of minimal magnitude. To attain these properties a fabric of relatively small thickness is provided, subject to the attainment of an adequate level of load bearing capacity and filament durability.

For optimum performance the elastomeric sheet materials of the invention are also fabricated according to a number of other geometrical criteria. These criteria provide not only the above mentioned properties in terms of thickness, compliance and stress-strain characteristics, but also the properties that relate to the promotion of desirable natural tissue layering upon the sheet material. Thus the sheet material is formed to provide a fibrous reticular face. Upon implantation, there is formed upon this face, because of its specific fibrous reticular nature, a satisfactory thin membranous deposit of endothelial cells, without an appreciable fibrous overgrowth of proteinaceous fibres such as collagen.

Accordingly, therefore, the present invention pro-

vides a flexible and extensible textile sheet material for cardiovascular and prosthetic implants, comprising a synthetic elastomer and having at least one fibrous reticular face.

- 5 Suitable elastomeric sheet materials may comprise flat woven or braided yarns forming distributed foramina of approximate maximum lateral aperture dimensions. Such foramina may be defined by the spacing between yarns or by the spacing between  
10 fibres within each yarn, or by both types of spacing. Other textile fabrication methods for producing a reticular fibrous sheet may be employed, for example knitting, flocking, needlong, tufting and spun bonding, preferably the foramina has a maximum  
15 lateral aperture dimension of 40 microns.

Additional features of this invention comprise certain geometrical and structural relationships which, when embodied in multifilament elastomeric yarns and fabrics, provide sheet materials that closely  
20 approximate the properties of natural heart valve leaflets and other natural tissues.

The invention will now be more fully described with reference to the following drawings, that are illustrative only.

- 25 In the drawings:

Figure 1 shows the main frame of the preferred form of the heart valve,

Figure 2 shows a fabric ribbon in the configuration formed by inserting it into the main frame between  
30 the rod portions of its legs.

Figure 3 shows the second frame.

Figure 4 is a top plan view of the frame shown in Figure 1.

Figure 5 is a top plan view of the frame shown in  
35 Figure 3.

Figure 6 shows the partially fabricated heart valve with the fabric inserted into the main frame and cut open preparatory to cementing thereto.

Figure 7 is a developed view of the partially constructed heart valve, corresponding to figure 6,  
40 Figure 8 is a cross sectional view taken on line 8-8 in Figure 7.

Figure 9 illustrates a flat braided fabric pattern.

Figure 10 is a stress-strain diagram comparing the  
45 characteristics of natural leaflet tissue prior art woven synthetic implants, and elastomeric sheet material according to this invention.

Figure 11 is a set of typical design graphs used for selection of the appropriate combinations of yarns  
50 per unit of fabric width, denier, and number of filaments per yarn, thus taking into account the filament densities and diameters to achieve the desired hole dimensions, that is, the aperture dimensions of the foramina.

Figures 1 to 8 show a preferred form of aortic trileaflet heart valve replacement. Referring to figure  
55 1, there is shown a main frame 22 comprising a single length of 0.1 cm diameter round polyp-ropylene rod bent into a form having three mutually equidistant, generally parallel legs 24, 26 and 28, each leg comprising a pair of rod portions slightly spaced apart, the rod portions being connected at one end and diverging at the other end. The diverging rod portions form three lobes 30, 32 and 34. The  
60 connected ends of the rod portions in each pair form

bights 36, 38 and 40. Figure 4 is a top plan view of the main frame 22.

A second frame 42 (figures 3 and 5) is formed of a single length of 0.1 cm diameter round polyp-  
70 ropylene rod bent into a form having three lobes 44, 46 and 48 generally congruent with the lobes 30, 32 and 34 so as to fit in close contact therewith as shown in figure 7.

The assembly is started by threading a ribbon 50  
75 of elastomeric sheet material of the type hereinafter described, in this case a plain woven elastomeric fabric, through the three pairs of rod portions so as to produce the configuration shown in figure 2. The yarns of the fabric are multifilament yarns. The  
80 frame 22 is shown in figure 1 in exploded relation to figure 2 for clarity of illustration. The upper selvage has no uncut yarns and forms the free edges 52, 54 and 56 of valve leaflets.

Thus a double layer of the fabric is passed through  
85 each pair of rod portions forming one of the legs 24, 26 and 28. It is necessary to attach the fabric firmly to these legs, and also to the connecting lobes 30, 32 and 34. To facilitate this attachment, the fabric is preferably cut lengthwise externally of each leg as  
90 shown in figure 6. adhesive such as polyurethane dissolved in tetrahydrofuran is applied to attach the fabric to each of the legs as follows. Flaps such as 58 and 60 are spread apart and the adhesive is applied at the external point of juncture of the flaps where  
95 they enter between the rod portions, in a continuous line extending between points a and b. The adhesive material reaches to the external surfaces of the frame by penetration through the fabric flaps along this line; that is, the adhesive contacts the rod  
100 portions of the frame only on their outer surfaces. The leaflets comprise only those portions of the fabric on the inside of the frame, and these portions are not penetrated by the adhesive. Thus local stiffening and resultant flex failure caused by such adhesive penet-  
105 ration is avoided.

The above method of adhesive application also distributes the stresses of flexure evenly along the margins of the leaflets, and avoids excessive stress concentrations. These margins are permitted to  
110 move upon each flexure over the rounded contours of the surfaces of the rod portions that are located on the inside of the frame, and that are not penetrated by the adhesive.

The attachment of the fabric to the lobes 30, 32  
115 and 34 is next accomplished by first placing the second frame 42 adjacent these lobes with the fabric pieces passing therebetween as shown in figures 7 and 8. Adhesive 61 is then applied through the fabric and to the surfaces of both the main frame 22 and the second frame 42, in a continuous line extending  
120 between the points b of the respective legs and connecting these three points. As in the previous step, the adhesive material preferably does not penetrate any portion of the leaflet material lying within the  
125 main frame 22, and remains out of contact with blood passing through the valve.

The foregoing steps essentially complete the fabrication of the leaflet portions of the valve. The remaining steps of fabrication are for the purpose of  
130 facilitating the suturing of the prosthesis within the

heart. The excess fabric available on the outside of the frame can be rolled and consolidated along the junction line between the main and second frames to provide attachment points for stitches during surgical insertion.

The frame material is preferably polypropylene, although other materials have also been employed with success. Polypropylene has excellent flex endurance and chemical stability, but is difficult to attach by adhesive to other materials. To facilitate adhesion, the main and second frames 22 and 42 may be encapsulated with polyurethane by multiple dip coating. The resulting encapsulated frame components have proved to demonstrate the desired characteristics of polypropylene without structural failures or breakdowns at the adhesive junctures. Another preferred material of construction for the frame is the cobalt alloy sold under the trade designation "Elgiloy".

Valves employing the elastomeric fabric described herein have been tested in an accelerated fatigue tester to assess their long term endurance characteristics. Fatigue failures so induced in prior art leaflet materials have generally occurred in the region of greatest fabric flexure, that is, along a line in each leaflet that is perpendicular to its free edge and substantially equidistant between the contiguous legs. The failures have generally occurred by breakdown of the filaments in the yarns running parallel to the free edge of the leaflets. As a means of providing greater fabric strength along the last mentioned lines, woven fabrics may be provided with a greater number of load bearing yarns in this direction. However there is a limit to the increase that is possible using a plain woven pattern without seriously disturbing the geometry of the fabric interstices.

An alternative fabric construction pattern having improved strength against such fatigue failure is illustrated in figure 9. The fabric shown is a flat braided ribbon 62 comprising 3 sets of elastomeric yarns, namely, a first diagonal set 64, a second diagonal set 66 and a longitudinal inlaid set 68. The yarns in each of the three sets are preferably multifilament untwisted yarns. The ribbon 62 is braided on a conventional flat braiding machine. It will be noted that each selvage has uncut yarns and one of those becomes the free edge of each leaflet. Thus fraying of the free edges of the leaflets is avoided as in the example described above. In this embodiment both of the sets 64 and 66 perform the load bearing function of a single set of yarns in the earlier described fabric. The result is that a greater number of yarns have a significant component of load bearing capacity oriented parallel to the free edge.

The fabric 62 of figure 9 is preferably formed by braiding the yarn sets 64 and 66 with inlaid longitudinal yarns 68 in a well-known manner, thus producing a type of triaxial fabric. Such flat braided fabrics have an additional advantage over conventionally woven fabrics, in that they are inherently highly extensible in the cross machine direction, that is, in the direction perpendicular to the yarns 68. Such fabrics make it possible to produce a two-way stretch characteristic.

In the foregoing description, woven and braided

fabrics have been described as embodied in replacement heart valve leaflets. However, many of the attributes of these fabrics as well as other textile sheet materials within the scope of this invention which have similar properties and are produced by such methods as knitting, flocking, needling, tufting spun bonding, etc, make them ideally suited for other biomedical applications as well. For example, vascular prosthesis, particularly those with small diameters, require a combination of good stretch characteristics and inherent biological inertness. In this case, fabrics essentially similar to any of those described herein may be woven, braided or otherwise fabricated in tubular form for use as conduits for flowing blood.

With either of the above tubular weave patterns, two-way stretch characteristics may be imparted to the vascular prosthetic devices. The radial compliance is particularly useful in avoiding stiffness mismatches at the boundaries between the existing artery and its synthetic replacement, particularly in small diameter arteries. A circumferential extension ratio of 1.5:1 is typically necessary in order to provide the proper match of properties, and this ratio has been provided by the above described tubular materials.

We turn next to a description of the preferred fabrics for use in the above described heart valve application and other prosthetic implant uses.

The preferred elastomers comprising the sheet materials of this invention are thermoplastic polyether esters prepared by transesterification from terephthalic acid, polytetramethylene ether glycol and 1,4 - butanediol. These copolyesters possess a two phase domain structure consisting of continuous and interpenetrating amorphous and crystalline regions. The amorphous elastomeric polyalkylene ether terephthalate soft segments contribute the elastomeric character to the polymer, whereas the crystalline tetramethylene terephthalate hard segments serve as thermally reversible tie points which are capable of holding the polymer chains together without the need for conventional covalent crosslinks. The synthesis of these copolymers is described in an article by G. K. Hoeschele, entitled "Segmented Polyether Ester Copolymers — A new generation of high performance Thermoplastic Elastomers", appearing in *Polymer Engineering and Science*, December 1974, Vo. 14, No. 12. In the practice of this invention it is preferred to select those copolymers having relatively larger amounts of the soft segments as compared with the hard segments and specific examples tested have included the copolymers sold under the trademark Hytrel 4056 by E. I. du Pont de Nemours & Company (hereinafter called "Hytrel"). These exhibit exceptional low temperature flexibility, and when fabricated as multifilament yarns of suitable denier they can be woven or braided to produce fabrics having the desired properties.

Other elastomers that can be similarly fabricated are within the purview of the invention. These include (1) polybutylene terephthalate, (2) a block polyester polyurethane copolymer sold under the trade mark "Pellethane" by Upjohn Company, (3)



thermoplastic silicone block polymer, and (4) a thermoplastic polyester elastomer sold under the trade mark "Arnitel" by Akzo Plastics.

The foregoing elastomers are extruded as filaments using multiple orifice spinnerets in a conventional manner, and low-twist multifilament yarns are formed. These yarns are then woven or braided to form the prosthetic fabric, or otherwise fabricated into textile sheet material having the mechanical and structural properties hereinafter described.

The advantages of elastomeric materials include their relatively low tensile moduli at low levels of stress, as shown by figure 10. This figure illustrates the tensile stress-strain characteristics of the natural heart valve leaflet tissue and two synthetic yarn materials. Stress is measured in grams of tension force per centimeter of leaflet or fabric width and strain in percent of original length. Curve 69 represents the natural leaflet characteristic in the circumferential direction. Curve 70 represents the natural leaflet characteristic in the radial direction. Curve 71 is representative of a fabric formed with an elastomer according to this invention and specifically represents a fabric woven of Hytrel yarn. Curve 72 represents a fabric formed with a polyester, namely polyethylene terephthalate which has been microcrimped after weaving in the manner described in the above mentioned U.S. Patent No. 4,191,218. By suitable microcrimping the fabric of the curve 72 may be made to exhibit a low initial modulus up to a strain of about 20 percent, above which the crimps are straightened and the modulus is much higher. Either of curves 71 or 72 shows a modulus which is low enough at low levels of strain to perform satisfactorily in a heart valve prosthesis. However as noted above, the low modulus properties are achieved in different ways.

As noted above the elastomeric materials of the invention may be fabricated in any of several ways to form textile sheets having the described properties. The following description, as applied to the plain orthogonally woven embodiment in a heart valve implant, is illustrative.

#### Thickness

An important criterion of the woven fabric is its thickness. Preferably, this should not exceed approximately 0.06 cm, the thickness of the natural heart valve leaflet. In addition, if the fabric is composed of yarns each having eight or more filaments, the level of twist imposes a criterion as shown by the following expression:

$$4d \leq t \leq (2n)d \quad (1)$$

where "d" is the filament diameter or the minimum lateral dimension of the filament cross section where the latter is not circular but oval or otherwise of flattened shape, "t" is the fabric thickness, and "n" is the number of filaments per yarn. As used throughout this description "thickness" refers to a dimension at right angles to the plane of the fabric. (A separate criterion imposing an upper limiting value 12d is explained below under the heading "Flexural Rigidity"). The basis for expression (1) as applied to twist may be understood by reference to the following discussion.

The expression "4d" represents the minimum

theoretical thickness of the woven fabric of multifilament yarns. This is attained when the warp and filling yarns are equally crimped during fabric production, as by weaving, and the yarns are sufficiently flattened by reason of having a low level of twist.

The crimp here referred to is termed "structural crimp" and is distinguished from the crimp described in said application Serial No. 901,085 which results from compaction and shrinking operations on the woven fabric. When the structural crimp is thus evenly distributed, the fabric thickness twice the thickness of a yarn, and the minimum theoretical thickness of a yarn having any degree of twist is twice the diameter of a filament, as required to accommodate the continuous filament relocation that is a necessary concomitant of the twisted structure.

On the other hand, if the yarns were highly twisted they would assume a more nearly circular shape, and the yarn thickness could be approximated by assuming that its cross section is a square made up of  $n_1$  rows of filaments with  $n_2$  filaments per row. In that case the fabric thickness would be approximated by the expression  $2n_1d$ . However by reducing the level of twist the fabric thickness can be reduced, and the use of thin fabrics confers several benefits. The flexural rigidity of the fabric is reduced, as are the bending stresses and strains in the material; the fabric weight per unit area is also reduced, thus minimizing the inertia-controlled response time of the leaflet during its opening and closing actions; and the diffusion of nutrient into subsequent tissue deposits is hindered to a minimum extent. Moreover the use of more-or-less balanced structural crimp is preferred because it gives a uniform surface contour and hence a tissue deposit with a more uniform thickness. For these reasons it is desired to impose a limit on the level of twist.

Defining the yarn cross-section "aspect ratio" as the ratio of the width of the yarn to its thickness, experience has indicated that acceptable fabric geometries are obtained for aspect ratios greater than 2.0. A yarn having this aspect ratio, comprising "n" filaments arranged in "a" rows has a thickness of "ad" and a width nd/a of twice that value, from which it may be derived that for a fabric of balanced structural crimp configuration, the upper limiting fabric thickness would be  $(2n)_1d$ . In general, fabrics with thicknesses lying in the lower and central portions of the range defined by expression (1) are preferred.

If there are fewer than eight filaments per yarn, including the case of monofilament yarns, expression (1) is not generally applied as a criterion because the level of twist in fabrics of balanced structural crimp is not important as long as the fabric thickness does not materially exceed the preferred face to face value of 0.06 cm previously mentioned.

An example of a suitable fabric according to expression (1) is an orthogonally woven fabric of identical warp and filling Hytrel yarns, there being 30 filaments per yarn each filament of 20.6 micron diameter. According to expression (1), the lower and upper limiting thickness of the fabric are 82 and 160 microns, respectively. The actual measured thick-

ness of the given fabric is 157 microns.

#### *Interfilament and Interyarn Hole Distribution*

As described in said U.S. patent 4,191,218 it is preferred to have the lateral dimensions of the foramina, holes or interstitial spacings between the yarns, or between the yarns and filaments, in the range of 20 to 40 microns. In particular, it is preferred that at least 50 percent of the superficial area of at least one face of the fabric contains a substantially uniform distribution of foramina having a maximum lateral aperture dimension of 40 microns. This imposes certain requirements upon "N", defined as the number of yarns per cm. of width in the fabric, as shown by the following expression:

$$\frac{1}{cb + bd} < N < \frac{1}{c + bd} \quad (2)$$

in which "c" is defined as the average hole diameter and preferably lies in the range between  $20 \times 10^{-4}$  and  $40 \times 10^{-4}$  cm, "d" is the filament diameter, and

"b" is the number of filaments per row in each yarn. The maximum value in equation (2) is determined by the case in which the only appreciable apertures through the fabric are the spaces between the yarns, the filaments in each yarn being in close side-by-side relationship. The minimum value is determined by the case in which the filaments of the yarns are separated to an extent sufficient to make the apertures between the yarns no greater than those between the respective filaments of each yarn.

When a fabric is to undergo fibre redistribution (spreading) in the finishing processes, then the minimum limiting value in expression (2) can be used as a guide to the type of fabric structure that can be manipulated from some starting configuration to the final desired geometry. When the fabric is intended for use in the as-woven consolidated configuration, with little or no filament redistribution the maximum value can be used to give the specification of the woven fabric. It is convenient in these latter cases to produce design graphs that embody the analytical relationship, cast in a form suitable for direct use. This is shown in figure 11, for example, which shows a series of graphs based on fabrics in which the yarns are in the consolidated, two layer configuration ( $a = 2$ ) with hole diameters of 30 microns, the filament density being 1.4 gm/ml. Similar graphs can be constructed for other yarns and also for other weave pattern geometries.

The graphs shown in broken lines in figure 11 show, for yarns of 30, 60, 90 and 120 denier, the value of "N" for each value of "n". The graphs shown in solid lines show, for yarns having 1, 1.5, 2, 3 and 4 denier per filament, the value "N" for each value of "n".

An example of a numerical calculation using the upper limit value in expression (2) is given below for the Hytrel fabric described above having 30 filaments per yarn, each filament of 20.6 microns. This fabric is designed to be used without any subsequent filament redistribution; therefore, all the effective interstitial holes are between yarns, which are themselves organised in approximate three-layer configurations ( $a = 3$ ). In this case,  $b = 30/3 = 10$ ;  $d = 20.6$  microns, and "N", the design threads per unit length for 30 micron holes is given by:

$$N = \frac{1}{30 \times 10^{-4} + 10 \times 20.6 \times 10^{-4}} = 42.4/\text{cm} \quad (3)$$

and fabric woven to this specification has a structure in which most of the interyarn spaces are approximately 30 microns in extent.

#### *Flexural Rigidity*

It has been pointed out that flexural rigidity of the fabric should be held to a minimum. A mathematical analysis of this property takes into account the Young's modulus "E" of the material, the second moment of area "I<sub>f</sub>" of the cross section of a fibre, and the effective number of rows of fibres in the yarn. The general formula for the flexural rigidity "G<sub>f</sub>" of a single fibre is

$$G_f = EI_f \quad (4)$$

For the case where the "n" fibres in a yarn are all completely free to move within the cross section the value G<sub>f</sub> in equation (4) multiplied by "n" would equal the flexural rigidity of a yarn. On the other hand, such complete freedom does not exist in a prosthetic implant because the fibres have a tissue deposited thereon and this may lead to a minimal to maximal increase in fibre restraint and stiffening of the fabric. In the worst case, if the restraint on the fibres were complete such that they were maintained at all times during bending in a configuration of "a" rows of "b" filaments per row with the neutral plane for bending being at the innermost row, it may be derived that the stiffness would be increased by the following factor f<sub>s</sub>:

$$f_s = \frac{2(2a^2 + 1)}{3} \quad (5)$$

over the value given above for the case where the fibres are completely free. Expression (5) shows that rigidity increases rapidly with increases in "a". Thus if "a" were "3", the maximum stiffening effect that could be imposed in theory upon the fibres would be approximately one order of magnitude over the completely free case. In practice, however, the tissue actually formed does not have the maximum stiffening effect, and it has been found that the value of "a" may be as high as six without an undesirable stiffening effect. As a more practical measure, it is preferred to establish as a criterion that the yarns have a minor axis (thickness) that is no more than six times the average filament diameter "d" producing a maximum fabric thickness of 12d.

Reverting to the previously given example of a Hytrel fabric having 30 filaments per yarn, each filament of 20.6 microns, the measured fabric thickness of 157 microns is well under the limit of  $12d \times 247.2$  microns established by the criterion for flexural rigidity.

#### *Tensile Compliance*

The "N" yarns per unit of width in the fabric, each yarn being composed of "n" filaments, must be such as to provide a tensile strength and modulus approximating those of the natural valve leaflet, described above. This places restraint on the available material and techniques. One centimeter width of artificial leaflet material will contain Nn filaments, using the nomenclature defined above, and will have an effective cross-sectional area of  $(Nn \pi d^2)/4$  sq.cm. If the tensile modulus of the fibre material is E gm/cm<sup>2</sup> then the modulus of the fabric in gm/cm width will

be  $E_t = (EN \pi d^2)/4$ . Most conventional textile materials have tensile moduli that fall in the range  $0.4$  to  $1.0 \times 10^6$  p.s.i., and are at least two orders of magnitude stiffer than is required to match the stiffness of the heart valve material at maximum stress, and several thousand times stiffer than the heart valve material at low stress levels. There are two useful approaches to the realization of low tensile modulus: in the first approach described in said copending application, excess filament length is introduced into the fabric in the form of crimp. In the second approach described in the present application low-modulus elastomeric materials are used as the fibre material, and the matching is achieved by means of

an overall reduction in the slope of the stress-strain characteristic. Using elastomeric materials of inherently low modulus, no crimping except for structural crimp resulting from fabric production itself is needed, and the fabric is geometrically simpler to model.

Several low modulus elastomeric materials, identified above, have been examined for their mechanical suitability for the leaflet application. These were spun into yarns with various filament diameters and deniers and their tensile behaviour was measured. Data on a selection of these materials is given in Table 1 below.

TABLE 1  
TENSILE PROPERTIES OF LOW MODULUS ELASTOMERIC YARNS

| (1)             | (2)         | (3)              | (4)                      | (5)                      |
|-----------------|-------------|------------------|--------------------------|--------------------------|
|                 |             |                  |                          | Initial Yarn             |
| Fabric Material | Yarn Denier | No. of Filaments | Filament Diameter, $\mu$ | Tensile Modulus (gm/den) |
| Pellethane      | 113         | 30               | 21.8                     | 0.09                     |
| Silicone        | 105         | 30               | 23.8                     | 1.14                     |
| Arnitel         | 81          | 30               | 18.7                     | 4.44                     |
| Hytrel          | 91          | 30               | 20.4                     | 0.70                     |
| Hytrel          | 91          | 30               | 19.5                     | 0.70                     |
| Hytrel          | 71          | 30               | 16.8                     | 0.63                     |
| PBT             | 210         | 30               | 26.8                     | 6.9                      |
| PBT             | 62          | 30               | 16.0                     | 7.4                      |

In Table 1, "Silicone" refers to the thermoplastic silicone block copolymer previously identified, and "PBT" refers to polybutylene terephthalate, also previously identified. The first listed Hytrel example having a filament diameter of 20.4 microns is the same yarn incorporated in the fabric example previously discussed, having 30 filaments per yarn, although the measurement in the fabric gave a somewhat higher filament diameter reading of 20.6 microns.

Fabrics woven to have desirable tissue reactions have holes that generally fall within the range 20 to 40 microns, as stated above. Preferably at least 50 percent of the superficial area of at least one face of the sheet material contains a substantially uniform distribution of foramina having a maximum lateral aperture dimension of 40 microns. This requirement is satisfied for the yarns described in Table 1 by incorporating them into fabrics with a yarn density "n" of approximately 40 yarns/cm.

The modulus of the natural leaflet material at the working load level is between approximately 1000 and 3600 gms/cm width. Any of the materials in Table 1, and others with similar tensile properties, are capable of being incorporated into fabrics that have both suitable geometric configurations and the proper tensile response, within a factor of two of these values.

Table 2 lists actual measurements of three woven fabrics each using the first listed Hytrel yarn in Table 1. The fabrics differed somewhat in the average number of yarns per centimeter and in processing, but all were satisfactory for use in prosthesis.

TABLE 2  
TENSILE PROPERTIES OF WOVEN  
HYTREL FABRICS

| Average Tensile Modulus Up to 150 gms/cm Width |                   |  |
|--|-------------------|--|
| Warp   | Filling           |  |
| Direction                                      | Direction         |  |
| $0.7 \times 10^3$                              | $0.6 \times 10^3$ |  |
| $0.5 \times 10^3$                              | $1.0 \times 10^3$ |  |
| $0.4 \times 10^3$                              | $0.6 \times 10^3$ |  |

Thus, either by suitable manipulation of the geometric form of the filaments and yarns, or by proper choices of filament tensile modulus, or by a combination of these techniques, it is possible to produce fabrics that have the desired combination of properties for heart valve leaflet applications, and also for other prosthetic and medical applications that demand approximately the same property combination. In general, a fabric having a modulus that does not exceed twice the circumferential tensile modulus of the heart valve leaflet material of 3600 grams/cm width, and is not less than half the radial tensile modulus of 1000 gm/cm width at the working load level of 150 gm/cm width will be adequate, and the mechanics of the valve suggest that the lower the modulus at extremely low levels of strain the better the performance will be. The preferred elastomeric multifilament yarns have an average tensile modulus up to a strain of 10 percent in the range of 0.05 to 5.0 grams per denier, the denier of the filaments being within the range between 0.5 and 20.

#### CLAIMS

1. A flexible and extensible textile sheet material for cardiovascular and prosthetic implants, comprising a synthetic elastomer and having at least one

fibrous reticular face.

2. The material of claim 1 containing distributed foramina with a maximum lateral aperture dimension of 40 microns.

5 3. The material of claim 2 in which the maximum lateral aperture dimensions of the foramina are defined by the interstitial spacing of fibres.

4. The material of claim 1, 2 or 3 having a maximum face to face thickness of about .06 cm.

10 5. The material of any one of claims 2 to 4 wherein at least 50 percent of the superficial area of at least one face thereof contains a substantially uniform distribution of foramina.

6. The material of any one of claims 1 to 5 comprising a segmented random block polyether ester copolymer in fibrous form, said copolymer further comprising crystallizable tetramethylene terephthalate hard segments and amorphous elastomeric polyalkylene ether terephthalate soft segments.

20 7. The material of claim 6 comprising a sheet of multi-filament yarns.

8. The material of claim 7 in which the yarns are woven.

9. The material of claim 7 in which the yarns are  
25 braided.

10. The material of any one of claims 7 to 9 comprising multifilament yarns having filaments of a denier between 0.5 and 20.

30 11. The material of any one of claims 7 to 10 comprising multifilament yarns having an average tensile modulus up to a strain of 10% which is between 0.05 and 5.0 grams per denier.

12. The material of any one of claims 7 to 11 wherein the number of yarns "N" per centimeter is  
35 given by

$$\frac{1}{cb + bd} < N < \frac{1}{c + bd}$$

where "c" is between  $20 \times 10^{-4}$  and  $40 \times 10^{-4}$  centimeter "d" is the filament diameter in centimeters,

40 "b" is the effective number of filaments per row in the yarn, and the effective number of rows of filaments in each yarn is between 2 and 6.

13. The material of any one of claims 7 to 12 comprising multifilament yarns and having a thick-

ness "t" given by

$$4d \leq t \leq (2n)d$$

where "d" is the filament diameter and "n" is the number of filaments per yarn and is greater than or equal to eight.

50 14. The material of any one of claims 1 to 13 having an average tensile modulus up to a load level of 150 grams per centimeter of width which is between 500 and 7200 grams per centimeter of width.

15. A method according to claims 12 or 13 comprising an orthogonally woven fabric formed from  
55 yarns containing 30 filaments/yarn, each filament having a diameter of 20.6 microns; a thickness of 15.7 microns, 42.4 threads per cm. and an interyarn spacing of 30 microns.

60 16. A material according to claim 1 substantially as herein described with reference to the drawings.

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A5R RAP

(56) Documents cited

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(58) Field of search

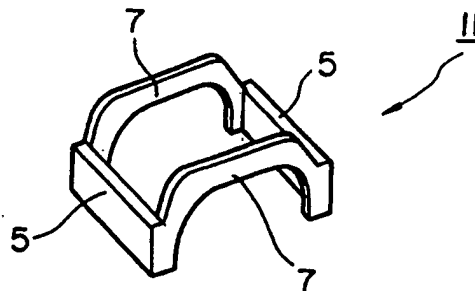
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INT CL<sup>5</sup> A61F 2/24

(54) Heart valve occluder supporter and a method of attachment thereof

(57) An occluder supporter (1) is installed to the valve of a mammalian heart for preventing valve reflux by stitching attachments (5) to the edges of the occluder. At least one bridging piece (7) spaces the attachments (5) to maintain the edges of the occluder in a mutually close position. The bridging piece(s) are provided between the attachments in such a way as to traverse the valve aperture. A pair of end bridging pieces (7 as shown) or a central bridging piece (7 Figure 4) may be provided. In further embodiments three bridging pieces (27 Figure 3) radiate from a central point to respective attachments (25) or form a triangular structure (32 Figure 9) with attachments (25) at the apices.

FIG. 1



GB 2 254 254 A

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FIG.1

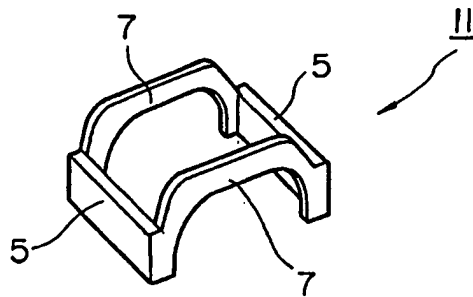


FIG.2

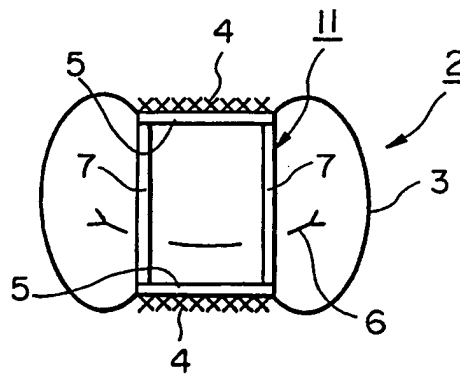
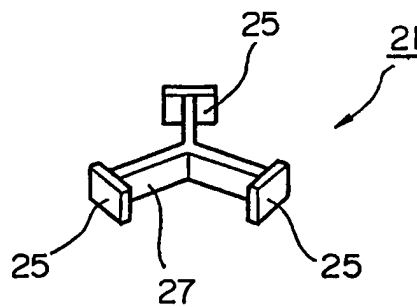


FIG.3



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FIG.4

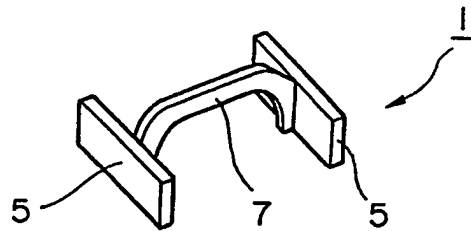


FIG.5

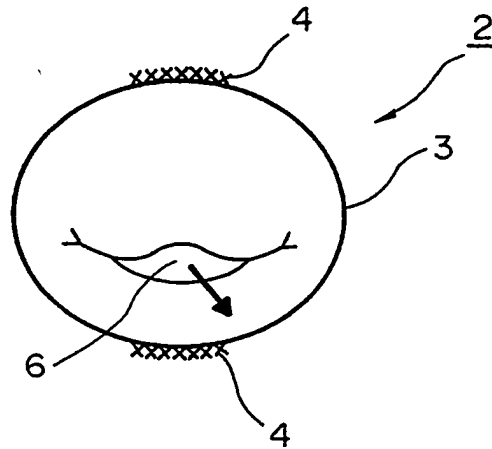
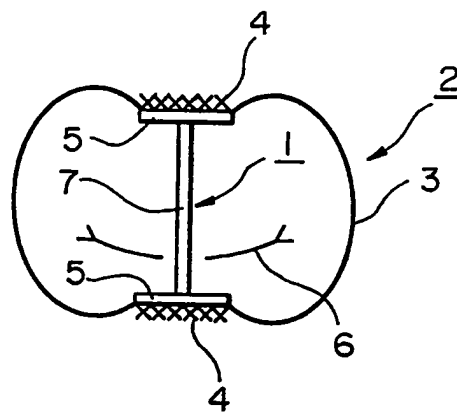


FIG.6



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FIG.7

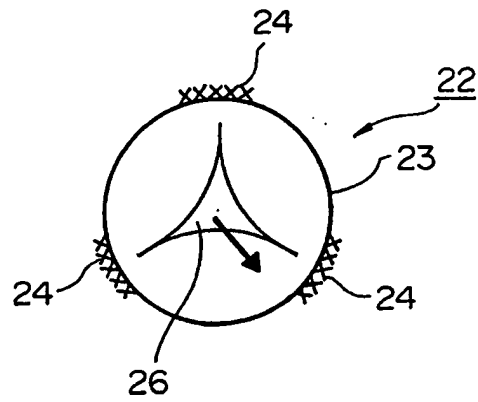


FIG.8

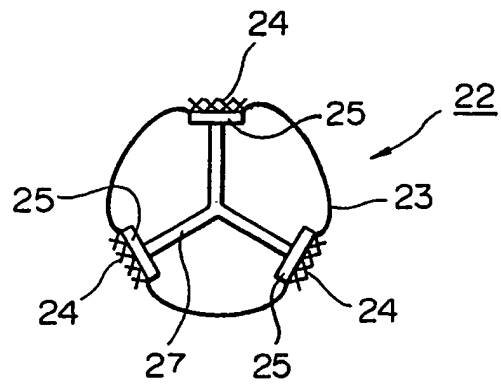


FIG.9

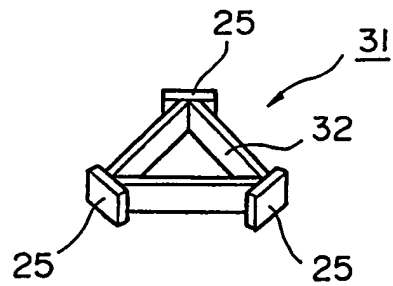
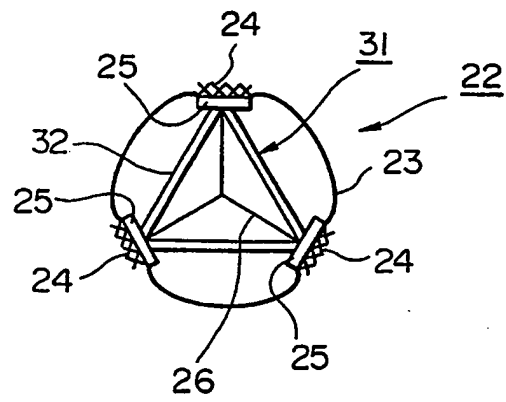




FIG.10



## OCCLUDER SUPPORTER AND A METHOD OF ATTACHMENT THEREOF

The present invention relates to a heart occluder supporter, and, in particular, to an occluder supporter which, through its provision to the valves of a mammalian heart, prevents blood from refluxing back through the valve.

There are four valves present in the mammalian heart. They are the tricuspid valve, the mitrial valve, the pulmonary valve, and the aortic valve.

The tricuspid valve is the valve located in the space between the right atrium and the right ventricle. The mitrial valve is the valve located in the space between the left atrium and the left ventricle. The pulmonary valve is the valve located in the space between the right ventricle and the pulmonary artery. The aortic valve is the valve located in the space between the left ventricle and the aorta. Of these, the tricuspid valve and the mitrial valve are called the ventricular valves.

A example of the mitrial valve and the aortic valve and an explanation of their function follows.

First, the case where the function of the mitrial valve is normal will be considered. During the contraction phase of the left ventricle, the mitrial valve is completely shut and the blood within the left ventricle is pumped to the aortic artery. During the relaxation phase of the left ventricle, the mitrial valve is completely open and blood flows into the left ventricle from the left atrium. If, for some reason, a situation occurs wherein the occluder of the bicuspid enlarges, or the chordae tendineae become extended or torn, the mitrial valve is unable to close completely, thereby resulting in a condition where a section of the valve aperture of the mitrial valve remains open. As a result, the opening and closing action of the mitrial valve is imperfect,

and a phenomena wherein blood refluxes from the left ventricle into the left atrium (valve reflux) occurs.

Next, the case where the function of the aortic valve is normal will be considered. During the contraction phase of the left ventricle, the aortic valve is completely open and the blood within the left ventricle is pumped to the aortic artery. During the relaxation phase of the left ventricle, the aortic valve is completely shut and blood flows into the left ventricle from the left atrium. As in the case given above for the mitral valve, if for some reason, a situation occurs wherein the occluder of the aortic valve enlarges, or the valve tip atrophies, the aortic valve is unable to close completely, thereby resulting in a condition wherein a section of the valve aperture of the aortic valve remains open. As a result, the opening and closing action of the valve is imperfect, and a phenomena wherein blood refluxes from the aortic artery to the left ventricle (valve reflux) occurs.

Valval transplant, valval suturing, and valvoplasty, are among the conventional methods of treatment to correct valve reflux.

Valval transplant is the method of excising the chordae tendineae of the valve tip and attaching an artificial valve or a donor valve to the remaining occluder. Valval suturing is the method of preserving the actual valve itself through suturing the occluder by sewing together one end of the occluder, or through setting the occluder by sewing on a ring to the occluder.

A variety of defects, however, are present in these methods.

For example, in valve substitution, the artificial or donor valve used is considerably expensive and has a short lifetime. Furthermore, in valve suturing and valvoplasty it is not possible to easily prevent valve reflux from occurring through the middle portion of the valve.

Accordingly, it is an object of the present invention to provide an occluder supporter which solves the problems encountered in the conventional technology and which can effectively prevent the occurrence of valve reflux.

In order to resolve the above problems, an occluder supporter is presented which can, by its installation in a mammalian heart valve, effectively prevent the occurrence of valve reflux. This occluder supporter has a plurality of stitched attachments which are attached to the edge of the occluder, and a support which supports the edges of the stitched attachments so that they are in a mutually close position and which is provided in the space between these stitched attachments so as to transect the valve aperture.

In this occluder supporter, the support holds the edges of the occluder in mutually close position while still providing a center space between them. As a result, even in the case where the occluder has enlarged, the valve can still shut completely and valve reflux can be prevented.

A device having a pair of stitched attachments, attached to the contralateral edges of the occluder, and a support, which holds and supports the contralateral edges in a mutually close position and is provided between the stitched attachments so as to transect the valve aperture, is preferable as an occluder supporter for provision to the ventricle side of the ventricle valve of the heart. Furthermore, it is preferable for the support to bend in such a way that it protrudes out on the ventricle side.

In an occluder supporter, the support holds and supports the contralateral edges of the occluder in a mutually close position while still providing for a center space between the contralateral edges. As a result, even in the event that the occluder has become enlarged or the chordae tendineae have become extended or torn, the aortic valve can still close completely during the contraction of the ventricle, preventing valve reflux. Furthermore, because the support bends in such a way as to protrude out on the atrium side, this support does not come in contact with the valve tip.

A supporter having three stitched attachments attached to each connecting part of the occluder and having a support which holds and supports these connecting part in a mutually close position and which is provided between the stitched attachments in such a way as to transect the valve aperture is preferable as an occluder supporter for provision to the aortic side of the aortic valve of the heart.

In this occluder supporter, the support holds and supports each connecting part of the occluder in a mutually close position, while still providing a center space between them. As a result, even in the event that the occluder has become enlarged or the valve tip has atrophied, the aortic valve can still close completely during the relaxation of the ventricle, preventing valve reflux.

Additionally, a method of attachment of the occluder supporter is proposed herein.

In this method of attachment of the occluder supporter, the support is disposed so as to create a condition wherein the edges of the occluder of a mammalian heart valve are held in a mutually close position, and so as to transect the valve aperture, and to these edges are attached stitched attachments. As a result, it is possible to easily and quickly attach the occluder supporter to the edge of the occluder.

The support is disposed in such a way as to transect the valve opening and to create a condition wherein the contralateral edges of the occluder of the ventricle side of the ventricle valve of a mammalian heart are held in a mutually close position, and the stitched attachments are attached to the aforementioned contralateral edges in order to attach the occluder supporter to the occluder of the atrium side of the ventricle valve. As a result, it is possible to easily and quickly attach the occluder supporter to the edge of the occluder.

The support is disposed so as to create a condition wherein the contralateral edges of the occluder of the atrium side of the ventricle valve of the mammalian heart are held in a mutually close position and so as to transect the valve

aperture, and the stitched attachments are attached onto the aforementioned contralateral edges in order to attach the occluder supporter to the occluder of the aortic side of the aortic valve. As a result, it is possible to quickly and easily attach the occluder supporter to the contralateral edges of the occluder of the atrial wall.

The following is a description of some specific embodiments of the invention, reference being made to the accompanying drawings in which: FIG.1 is a perspective diagram showing the occluder supporter of the first embodiment of the present invention:

FIG.2 is a schematic diagram showing the bicuspid valve as seen from the atrial wall of the heart.

FIG.3 is a frontal diagram showing the attachment of the occluder supporter of the first embodiment of the present invention to the bicuspid valve which is shown as seen from the atrial side.

FIG.4 is a perspective diagram showing a variation of the occluder supporter of the first embodiment of the present invention.

FIG.5 is a frontal diagram showing the attachment of a variation on the occluder supporter of the first embodiment of the present invention to the bicuspid which is shown as seen from the atrial side.

FIG.6 is a perspective diagram showing the occluder supporter of the second embodiment of the present invention.

FIG.7 is a schematic diagram showing the aortic valve as seen from the aortic side of the heart.

FIG.8 is a frontal diagram showing the attachment of the occluder supporter of the second embodiment of the present invention to the aortic valve which is shown as seen from the aortic side.

FIG.9 is a perspective view showing a variation of the occluder supporter of the second embodiment of the present invention.

FIG.10 is a frontal diagram showing the attachment of a variation on the occluder supporter of the second embodiment of the present invention to the aortic valve which is shown as seen from the aortic side.

An explanation of each of the preferred embodiments of the present invention, with reference to the figures, follows below.

#### First Embodiment

FIG.1 is a perspective view diagram showing occluder supporter 1 of the first embodiment of the present invention.

This occluder supporter is provided to the left atrium side of the bicuspid valve and is a device for preventing valve reflux. Occluder supporter 1 is comprised from stitched attachments 5,5 and support 7.

Stitched attachments 5,5 are a pair of tabular plate bodies sewed on to contralateral edges 4,4 of occluder 3 of the bicuspid valve 2 shown in FIG.2. Furthermore, support 7 is provided so as to traverse valve aperture 6 in the space between stitched attachments 5,5 and hold and support contra-lateral edges 4,4 in a mutually close position. Support 7 bends in an arc shape and has a streamlined rhabdome cross section.

The base substance of stitched attachments 5,5 and support 7 is comprised from Titanium (Ti), and the surfaces are coated with alumina ( $Al_2O_3$ ), a ceramic which is antithrombotic and antihemolytic. Furthermore, the area of attachment between occluder 3 and stitched attachments 5,5 is covered with poly tetra fluoroethylene, an organic material which is antithrombotic and antihemolytic character.

Furthermore, in addition to Titanium, other noncorrosive metal such as Titanium alloy, stainless steel or the like may be appropriately used as the material for stitched attachments 5,5 and support 7. Additionally, in addition to alumina, other organically appropriate antithrombotic or antihemolytic ceramic such as hydroxyapatite can be used as the ceramic. And, in addition to poly tetra fluoroethylene, other antithrombotic, antihemolytic polyester and the like may be used as the organic material.

As is shown in FIG. 3, in order to attach occluder supporter 1 to occluder 3 of bicuspid valve 2, support 7 is

disposed so as to hold contralateral edges 4,4 of occluder 3 in a mutually close position and so as to traverse valve aperture 6. Stitched attachments 5,5 are attached to each contralateral edge 4,4.

Because the above occluder supporter 1 comprises stitched attachments 5,5 and support 7, it is possible for the mitrial valve to close completely during the contraction phase of the left ventricle. Accordingly, it is possible to prevent valve reflux.

Because support 7 is a rhabdome shape which bends in such a way as to protrude out on the left atrial side, it is possible to reduce the resistance offered to blood flow. Moreover, support 7 does not contact with the valve tip.

Because the surfaces of stitched attachments 5,5 and support 7 are covered with an aluminum material having antithrombotic and antihemolytic properties, it is possible to improve the biocompatibility of stitched attachments 5,5 and support 7. Furthermore, it is possible to markedly improve the durability of the occluder supporter without the occluder supporter itself becoming a cause of thrombosis or hemolysis.

In stitched attachments 5,5, because the area of attachment to occluder 3 is covered with poly tetra fluoroethylene, an organic material which is antithrombotic and antihemolytic, it is possible to improve the organic compatibility of stitched attachments 5,5, and to easily carry out the connection of stitched attachments 5,5 to occluder 3.

According to the method for attachment of occluder supporter 1, because support 7 is disposed so as to hold contralateral sides 4,4 of occluder 3 in a close position and so as to transect valve aperture 6, and because stitched attachments 5,5 are attached to each contralateral edge 4,4, occluder supporter 1 can be easily and quickly attached to contralateral edges 4,4 of occluder 3 of the atrium side.

FIG.4 is a perspective view showing a variation on the embodiment of occluder supporter 1.



In occluder supporter 11, two supports 7,7 are provided in parallel to stitched attachments 5,5.

As shown in FIG.5, occluder supporter 11 can be attached to occluder 3 of mitral valve 2 by the same method as employed for occluder supporter 1.

In occluder supporter 11 and in its method of attachment, the same effects and functions are performed as by occluder supporter 1 and in its method of attachment. Additionally, because two supports 7,7 are utilized, it is possible to improve the strength of the device over that of occluder supporter 1.

#### Second Embodiment

FIG.6 is a perspective view showing occluder supporter 21 of the second embodiment of the present invention. Occluder supporter 21 is a device which is installed to the aortic side of the aortic valve in order to prevent valve reflux. Occluder supporter 21 comprises three stitched attachments 25 and a support 27.

Stitched attachments 25 are of a tabular form and are sewn on to each connecting part 24 of occluder 23 of the aortic valve 22 shown in FIG 7. Furthermore, support 27 is installed between stitched attachments 25 in such a way as to transect valve aperture 26, and is of a tripartite rhabdome shape which supports and holds each connecting part 24 in a close position.

The base substance of stitched attachments 25 and support 27 is formed from Titanium. The surfaces of stitched attachments 25 and support 27 are coated with an antithrombotic, antihemolytic ceramic alumina. The antithrombotic, antihemolytic alumina material covers the surface of stitched attachments 25 and support 27. Additionally, the area of attachment of stitched attachment 25 to occluder 23 is covered by a poly tetra fluoroethylene, an organic material which is antithrombotic and antihemolytic.

In addition to Titanium alloy, other appropriate noncorrosive, antithrombotic, antihemolytic metal, such as stainless steel, may be used as the material for stitched

attachments 25 and support 27. Furthermore, in addition to alumina, other biologically compatible, antithrombotic, antihemolytic ceramic, such as hydroxyapatite, may be used as the ceramic. And in addition to poly tetra fluoroethylene, other appropriate antithrombotic, antihemolytic polyester may be used as the organic material.

As shown in FIG.8, support 27 is disposed so as to hold each connecting part 24 of occluder 23 in a mutually close position and so as to transect valve aperture 26, and stitched attachments 25 are sewn on corresponding to each connecting part 24.

Because occluder supporter 21 comprises stitched attachments 25 and support 27 it is possible to create a condition wherein the aortic valve 22 closes completely during the relaxation phase of the left ventricle, accordingly preventing valve reflux.

Because support 27 is a tripartite rhabdome shape, it is possible to improve the strength of support 27 while at the same time reduce the resistance offered with respect to blood flow.

Because the surfaces of stitched attachments 25 and support 27 are covered with an antithrombotic, antihemolytic alumina, it is possible to improve biologic compatibility thus ensuring that occluder supporter 21 itself does not become a cause of thrombosis or hemolysis, and to markedly improve the durability of stitched attachments 25 and support 27.

In stitched attachments 25, because the area of attachment to occluder 23 is covered with a poly tetra fluoroethylene, an antithrombotic, antihemolytic organic material, it is possible to improve the biological compatibility of stitched attachments 25 and to easily carry out the attachment of stitched attachments 25 to occluder 23.

According to the method of attachment of occluder supporter 21, support 27 is disposed so as to hold each connecting part 24 of occluder 23 in a mutually close position and so as to transect valve aperture 26, with stitched attachments 25 being attached corresponding to these

connecting parts 24. For this reason, it is possible to easily and quickly attach occluder supporter 21 to each connecting part 24 of occluder 23.

FIG.9 is a perspective diagram showing a variation on the preferred embodiment of occluder supporter 21.

Occluder 31 is formed from the provision of a triangular rhabdome shaped support 32 in the space between stitched attachments 25, and from the fixing of the top section of support 32 to each stitched attachment 25.

As shown is FIG. 10, occluder supporter 31 can be attached to occluder 23 of the aortic valve by the same method as employed for occluder supporter 21.

In occluder supporter 31 and in its method of attachment, the same effects and functions can be performed as by occluder supporter 21 and in its method of attachment. Furthermore, because support 32 is a triangular rhabdome shape it is possible to improve the strength of support 32 while at the same time to decrease the resistance offered to blood flow.

## CLAIMS

1. An occluder supporter which is provided to the heart valve of a mammalian heart in order to prevent the occurrence of valve reflux, the characterising feature of said occluder supporter being the provision of a plurality of stitched attachments which are sewn on to the edge of the occluder and with a support which is provided between the said stitched attachments in such a way as to traverse the valve aperture, and said support supporting and maintaining the edges of said stitched attachments in a mutually close position.

2. An occluder supporter which is provided to the atrial side of the ventricle valve of a mammalian heart in order to prevent the occurrence of valve reflux, the characterising feature of said occluder supporter being the provision of a pair of stitched attachments which are sewn onto the contralateral sides of the occluder and with a support which is provided between the said stitched attachments in such a way as to traverse the valve aperture, said support supporting and maintaining said contralateral edges in a mutually close position, and said support bending in such a way as to protrude out on the atrial side.

3. An occluder supporter which is provided to the aortic side of the aortic valve of a mammalian heart in order to prevent the occurrence of valve reflux, the characterising feature of said occluder supporter being the provision of at least 3 stitched attachments which are sewn onto each connecting portion of the occluder, and with a support which is provided between the said stitched attachments in such a way as to traverse the valve aperture, said support

supporting and maintaining each connecting portion in a mutually close position.

4. An occluder supporter as claimed in Claim 2  
5 or Claim 3, wherein the support has a rhaddome shape.

5. An occulder supporter as claimed in Claim 2  
or Claim 3, wherein the support has a stream-lined  
cross section.

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6. An occulder supporter as claimed in Claim 2,  
wherein a plurality of supports is provided in  
parallel in the space between said stitched  
attachments.

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7. An occulder supporter as claimed in Claim 3,  
wherein the support is of a tripartite rhabdome shape  
and said stitched attachments are fixed respectively  
to each edge of said support.

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8. An occluder supporter as claimed in Claim 3,  
wherein the support is of triangular shape and said  
stitched attachments are fixed respectively to each  
apex of said support.

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9. An occluder supporter as claimed in any of  
Claims 2 to 8, wherein the stitched attachments have a  
width greater than that of said support.

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10. An occluder supporter as claimed in any of  
Claims 2 to 9, wherein said occulder supporter is  
formed from antithrombotic and antihemolytic material.

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11. An occluder supporter as claimed in Claim 10,  
wherein at least the surface of said occluder  
supporter is covered by a ceramic material.

12. An occluder supporter as claimed in Claim 11, wherein the principle component of said ceramic is alumina.

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13. An occluder supporter as claimed in Claim 10, wherein at least the area of attachment between said stitched attachments and said occluder is covered by an antithrombotic, antihemolytic organic material.

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14. An occluder supporter as claimed in Claim 13, wherein the principle component of said organic material is one of either polytetrafluoroethylene or polyester.

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15. A method of attachment of an occluder supporter as claimed in any of Claims 1 to 14, wherein said support supports and maintains the edges of the occluder of a mammalian heart valve in a mutually close position, said support being disposed so as to traverse the aperture of said valve, and wherein said stitched attachments are sewn on to said edges of said occluder.

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16. A method of attachment of an occluder supporter as claimed in any of Claims 1 to 14, wherein said support supports and maintains the contralateral edges of the occluder on the atrial side of the ventricle valve of a mammalian heart in a mutually close position, said support being disposed so as to traverse the aperture of said valve, and wherein said stitched attachments are sewn on to said contralateral edges of said occluder.

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17. A method of attachment of an occluder supporter as claimed in any of Claims 1 to 14, wherein

said support supports and maintains each connecting part of said occluder on the aortic side of the aortic valve of a mammalian heart in a mutually close position, said support being disposed so as to  
5 traverse the aperture of said valve, and wherein said stitched attachments are sewn on to said connecting parts of said occluder.

18. An occluder supporter for the heart valve of  
10 a mammalian heart substantially as described with reference to and as illustrated in Figures 1 to 3 or Figures 1 to 3 as modified by Figure 4 or Figure 5 of the accompanying drawings.

19. An occluder supporter for the heart valve of  
15 a mammalian heart substantially as described with reference to and as illustrated in Figures 6 to 8 or Figures 6 to 8 as modified by Figure 9 or Figure 10 of the accompanying drawings.  
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(i) UK Cl (Edition K ) A5R (RAP)

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Databases (see over)

(i) UK Patent Office

(ii)

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Documents considered relevant following a search in respect of claims

1 TO 14, 18, 19

| Category<br>(see over) | Identity of document and relevant passages                            | Relevant to<br>claim(s) |
|------------------------|---|-------------------------|
| A                      | GB 1293014 (RHONE-POULENC S.A.) - see eg<br>drawings                  |                         |
| A                      | EP 0257874 A (BAXTER TRAUENOL LABORATORIES,<br>INC) - see eg Figure 1 |                         |



| Category | Identity of document and relevant passages | Relevant to claim(s) |
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